The relationship between frequency specific auditory brainstem response thresholds and behavioural audiometric thresholds in hearing impaired adults.

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ABSTRACT

Auditory brainstem response (ABR) thresholds to masked tone pip stimuli, were obtained for 3 groups of hearing impaired subjects. Using high pass noise at 500 Hz and notched noise at 1,2, and 4 kHz, ABR thresholds in low frequency, high frequency, and flat cochlear losses were compared to conventional pure tone audiometric thresholds. A strong positive relationship, with no systematic effects of either impairment group or frequency, was found between ABR (re: laboratory norms) and behavioural (re: 0 dB HL) threshold elevation, though absolute ABR thresholds (dB peSPL) at 500 Hz were significantly higher than those at other frequencies. Regression analyses of the data indicated a strong but variable linear relationship between ABR and behavioural measures. The results of this study indicate that frequency specific ABR testing can provide an approximation of both the degree and the configuration of cochlear hearing losses in adults. Further refinements of both the testing and judging procedures are needed however, to reduce the variation evident in our results, and thus achieve the accuracy required for most clinical applications.

ACKNOWLEDGEMENTS

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INTRODUCTION

The auditory brainstem response (ABR) is a far field recording of synchronised electrical impulses from the auditory brainstem. It is now used widely in the audiometric test battery, both diagnostically to check the integrity of the auditory pathways and as an objective measure of hearing sensitivity (Gorga & Worthington, 1983). Because of its ease of recording, high replicability and the fact that it does not depend on behavioural responses from the patient, it serves as a unique tool for evaluating both cochlear and brainstem function (Kaga et al, 1986).

First described in 1971 by Jewett (Jewett & Williston, 1971), its applications as an objective hearing test include testing the very young and other difficult to test populations such as the multiple handicapped and uncooperative (Weber, 1987). More recently it has been used with medicolegal patients in, for example, the assessment of compensation cases (Hyde et al, 1986). Also within the clinical setting, it has a possible application in selecting the frequency response and output amplitude characteristics of hearing aids in difficult to test populations. (Gorga & Worthington 1983, Mahoney 1985).

Morphologically, the ABR consists of five to seven wave peaks that can be reliably identified in the first 10 milliseconds

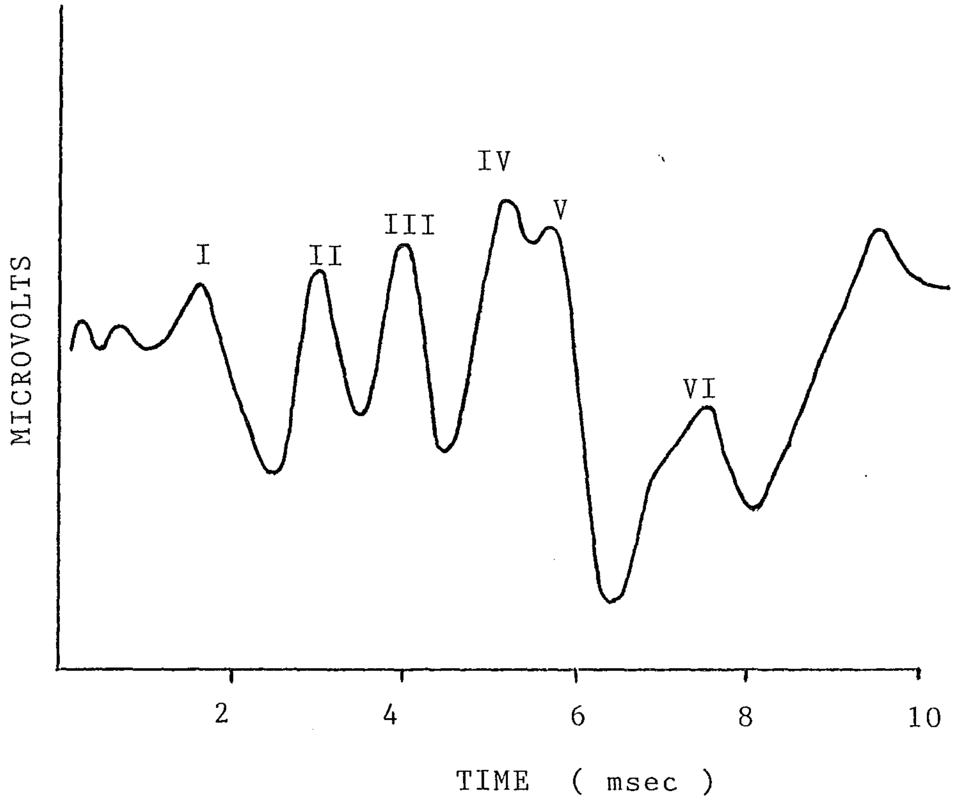


FIGURE 1. A normal ABR from an adult male (vertex-positive) with the successive peaks conventionally numbered as shown.

(from Schwartz & Berry, 1985, p.66)

after the evoking stimulus. (See Figure 1). Jewett and Williston (1971) developed a nomenclature which has been adopted as a standard to identify these waves using the Roman numerals I-VII. The absolute latency of these peaks, inter-peak latencies, interaural differences and other parameters are measured in the diagnostic applications of the ABR.

To help in the assessment of hearing status, the relationship between stimulus intensity and response amplitude has been investigated (Jacobson, 1985). Wave V, the most robust of the seven waves (Hawes & Greenburg, 1981), can be tracked via its systematic increase in latency and decrease in amplitude in response to decreased stimulus intensity (Starr et al., 1975, Jewett & Williston, 1971). The accuracy of the technique in predicting hearing thresholds depends on three main sets of variables: SUBJECT, STIMULUS, AND RECORDING FACTORS.

1) SUBJECT FACTORS

a) AGE: The morphology of the ABR varies with the age of the subject. Infant ABRs are usually characterised by the presence of waves I, III and V while the latency of wave V decreases with maturation through to the second year of life (Hecox & Galambos, 1974). This is thought to reflect the maturation of the auditory system and poses questions as to the validity of relating normative adult data to neonate and infant populations. Indeed, most clinics have separate norms for adult and infant patients. Some debate currently exists also over possible aging effects in geriatric populations (Schwartz & Berry, 1985). Thus the ABR is, to some degree, dependent on the age of the subject. The present study used adult subjects as adult ABRs are widely accepted as being stable with age.

b) SEX: Females have been found to have shorter wave V latencies than males, however, the amplitudes of the responses are similar (Rosenhamer et al., 1980). Hence, in threshold seeking techniques, where only the response amplitude is examined, no correction need be made for the sex of the patient.

c) STATE: From early work done by Jewett in 1971, the ABR has been shown to be independent of the subject's level of arousal or attention. Starr and Achor (1975) found that even patients with coma induced by metabolic or toxic factors still had recordable ABRs with normal latencies and amplitudes. Sedation is known to improve the quality of many evoked potential recordings by significantly improving the signal to noise ratios involved. Adams et al (1985) have shown that sedation with diazepam, for example, had no effect on ABR amplitudes in response to clicks. This is a particularly useful finding because the ABR is vulnerable to electrical masking by movement artefacts, particularly muscle action potentials in very young, unco-operative or unwell patients (Starr & Achor, 1975). Yung (1985) looked at whether routine sedation was necessary in ABR testing, and concluded that with relaxed and co-operative adults the sensitivity of the test was unchanged. Hence, the benefits of sedating patients clinically is restricted to infants and 'difficult to test' patients.

2) STIMULUS FACTORS

Before looking at the stimuli used in frequency specific testing, it is useful to consider briefly the mechanics by which frequency analysis is achieved in the cochlea - specifically via the travelling wave which occurs in the cochlea in response to sound, due to the physical properties of the basilar membrane and cochlear fluids. According to von Bekesy (1960), a travelling wave moves from the base to the apex of the cochlea and maximally displaces different regions of the basilar membrane depending on the frequency of the stimulus. This process produces place coding: sound frequency is coded by place on the basilar membrane. Stapells et al (1985) point out three main corollaries of the PLACE theory.

1. The time taken for the wave to reach the apex of the cochlea means that low frequency responses are initiated later than high frequency responses.

2. The asymmetry of the travelling wave means that low frequency sounds activate both basal (high frequency) and apical (low frequency) regions, while high frequency sounds only activate basal regions.

3. Progressively greater areas of the basilar membrane are activated as the wave moves from base to apex and, therefore, neural synchronity is greater in the basal areas than in the apical areas (Stapells et al., 1985).

As will be seen in the following discussion, each of these points is important to the theory behind frequency specific ABR testing.

The usual stimulus for evoking an ABR is the unfiltered click, generated by feeding a brief rectangular electric impulse to the earphone (Arlinger, 1981). Although ideal for elicting a synchronous neural response, the click - because of its rapid onset - has a flat frequency spectrum. This flat spectrum means that energy spreads over a wide range of frequencies causing activation of the entire length of the basilar membrane. Such a stimulus is therefore inadequate for predicting sharply sloping or notched hearing losses because responses will be initiated by frequencies other than those with elevated thresholds. Since the ultimate goal of such a test should be to reproduce accurately the fine details of the audiometric contour (Hyde et al., 1986), the failure of the click to be frequency specific makes its use as a predictive stimulus limited. As neural synchronicity is greatest in the basal (high frequency) region of the cochlea (Kiang 1965), the click evoked ABR has been generally accepted as best reflecting hearing sensitivity in the 2 - 4 kHz region (Davis & Hirsh 1976).

Frequency specific ABR research has been pursued in an attempt to fulfil the clinical need to estimate hearing-thresholds below 2 kHz. The main difficulty encountered in trying to achieve accurate low frequency hearing estimates from the ABR lies in designing a narrow band stimulus which provides sufficient low frequency specificity and yet is still rapid enough to elicit the well synchronised neural responses necessary for a clear ABR tracing (Borg, 1981). The problems of evaluating lower frequency (<2 kHz) hearing can be related to the decreased velocity of the travelling wave in more apical regions on the cochlea. The reduced synchrony of apical neural activity results in less clearly defined peaks in the response (Sohmer & Kinarti, 1984). It is an established fact that the amplitude of the ABR is dependent on the first few milliseconds of the evoking stimulus and that ABRs are determined primarily by neural activity which occurs at stimulus onset (Suzuki & Horiuchi 1981, Hecox 1976).

In summary, then, a stimulus used for a frequency specific ABR has to meet the demands of having:

a) sufficiently rapid onset to elicit a clear ABR

b) sufficient duration to minimise energy spread and thus provide frequency specificity. (Stapells et al, 1985).

Two stimuli which have been evaluated are the filtered click and the tone burst. A filtered click is achieved by passing a rectangular pulse through a bandpass filter with a set centre frequency and bandwidth. (See Figure 2 (b)). Arlinger (1981) states that if the frequency response of the transducer (ie. headphone) is reasonably flat, the acoustic signal produced will be a good approximation of the filtered electrical signal. Gibson (1978) however warns that analysis of the onset of the acoustic waveform often reveals high frequency transients associated with transducer resonances.

(a) (b) (c)

FIGURE 2. Examples of brief stimuli used in ABR testing.

a) the unfiltered (broadband) click.

b) the filtered click

c) the tone burst (from Arlinger 1981, p. 42)

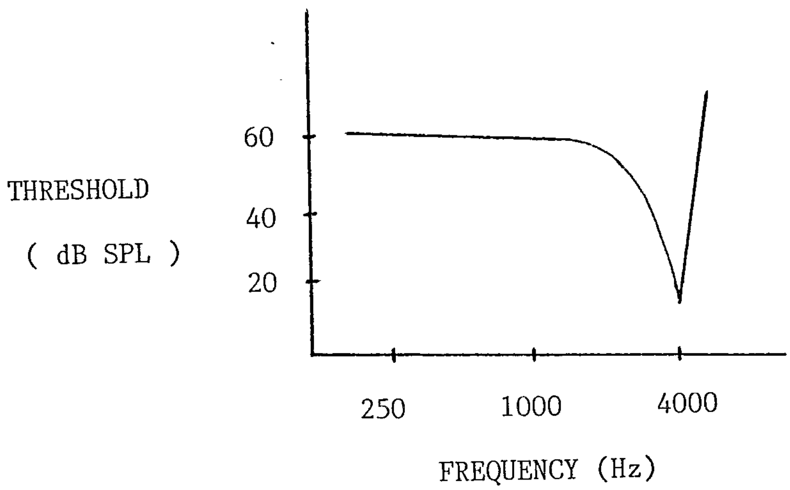


FIGURE 3. Tuning Curve for an auditory nerve fibre with a 4000Hz characteristic frequency. Note that it will respond to a 500Hz stimulus at 60 dB SPL intensity. (from Gorga et al., 1983, p.354)

The tone burst, or tone pip is shown in Figure 2 (c) and is a brief tone pulse with a controlled acoustic envelope (rise - plateau - fall times). Increasing the duration of these parameters narrows the frequency spectrum but at the same time makes the onset very gradual and, therefore, also decreases the clarity of the resultant ABR (Suzuki & Horiuchi, 1981). This finding was also reported by Bergholtz (1981) who found that ABR responses especially to low frequency stimuli such as the filtered click and tone burst, were comparatively small. He cited less neural synchronisation, associated with the increased rise times involved, as the probable cause. Hence, with both these stimuli, the fundamental problems of the onset / duration dilemma remains. In order to maintain a clear response through rapid onset, some loss of frequency selectivity through "spectral splatter" is necessary. Picton et al (1981) provided further evidence of the consequences of this dilemma, reporting that tracings from frequency specific stimuli with sufficiently rapid onsets to elicit a clear ABR were dominated by activity initiated more basally than where the nominal frequency of the stimulus should maximally stimulate. The implication of this energy spread clinically is that threshold predictions based on these "earlier" wave V components, may lead to underestimation of actual thresholds at a given frequency.

MASKING:

In order to overcome the spread of energy over the frequency spectrum when brief tonal or narrowband signals are used, various masking paradigms have been employed to eliminate the contribution of activity caused by 'spectral splatter'. The two assumptions made when masking is used are that:

1) the noise effectively masks responses from regions of the cochlea remote from that under examination by desynchronising neural activity from these regions.

2) the noise does not alter the responses from those regions of the cochlea being studied. (Gorga et al., 1983 p. 357).

Using masking to "block out" responses from "unwanted" areas of the cochlea provides a place specific method of threshold evaluation. Stapells et al (1985) pointed out that a patient with a "place specific" hearing loss, with no cochlea hair cells after the basal turn, may only have a mild to moderate low frequency loss on the audiogram yet on ABR testing a severe loss would be present. This is because pure tone low frequency stimuli, even when devoid of spectral splatter, will, at high intensities, still be picked up by high frequency auditory nerve fibres. (See Figure 3) In such a case, the "place specific" technique would provide a more accurate indication of cochlear status than conventional pure tone audiometry, though actual behavioral hearing thresholds would be overestimated. Thus, the various masking paradigms used, attempt to control the place of excitation and as such can be considered place specific tests (Hyde, 1985).

a) <u>DERIVED BAND ANALYSIS:</u>

The differential high pass masking paradigm or derived band technique is based on broad band click stimulation, and employs various levels of highpass masking noise presented with a click, to elicit a response supposedly from the unmasked region of the basilar membrane. Frequency specific responses are obtained by a series of mathematical subtractions and derivations (Don & Eggermont, 1978). Evidence both supportive and unsupportive exists for this technique's clinical accuracy. For example, Don et al.(1979) reported close correlation between behavioural and derived band ABR thresholds in normal and hearing impaired subjects. Laukli and Mair (1986) concluded, however, that the lack of low frequency energy in the click stimulus made the test unsuitable for evaluation of audiometric thresholds below 1 kHz. Sohmer & Kinarti (1984) felt that the derived band technique required further validation in hearing impaired subjects and questioned its validity at low frequencies even in normal subjects. In addition to this theoretical criticism, there is the practical disadvantage that it requires complex equipment and is time consuming in data acquisition and analysis (Arlinger, 1981, Gorga et al., 1983).

b) TONE BURSTS:

Recently, the tone burst has become the most commonly researched stimulus in ABR audiometry, and the envelope parameters of these bursts have been the subject of much debate. Stapells et al (1985) proposed a 'compromise' stimulus in an attempt to solve the dilemma of creating both a rapid onset yet still frequency specific stimuli. They suggested a tone burst with a 2 cycle rise/fall time and a 1 cycle plateau. The resultant energy spectra of this "2-1-2" stimulus is uniform regardless of the centre frequency being used. It also displays side lobes which are approximately 30 dB below the nominal frequency (Davis et al., 1984).

Numerous studies have also concluded that the brief tone burst with constant rise-plateau-fall parameters across frequencies is the most suitable stimulus for the ABR analysis of hearing thresholds (Davis & Hirsh, 1979, Hawes & Greenburg, 1981). Takagi et al (1985) reported that the ABR consists of two main components originating from different areas of the auditory system. A "fast" component was thought to be generated by synchronised action potentials while a "slow" component was generated by the brainstem nuclei. Finding that for low frequency stimuli the ABR was dominated by "slow" component activity, they concluded that this component was the most useful index in the ABR for the estimation of hearing threshold (Takagi et al., 1985, p.79). They also found that using constant period envelopes, such as the "2-1-2" tone, minimised inter-frequency differences in the amplitude of the slow component.

Other investigators (Beattie & Boyd, 1985, McGee & Clemis, 1980, Hayes & Jerger, 1982) have employed a constant rise/fall time in milliseconds rather than using frequency dependent periods. The advantage of the constant rise time tone is that once the travelling wave delays have been accounted for, the initiation of the auditory impulse occurs at an approximately equivalent time for each frequency (Stapells et al., 1985).

c) TONES AND HIGH PASS NOISE:

High pass masking noise in association with tone pips was used by Davis and Hirsh (1976). Using unmasked tone pips they found that, at higher stimulus levels, an "apical" response to low frequency stimuli could be obscured by an earlier and larger response initiated in the basal turn. This "basal" response could be eliminated by presenting high pass filtered masking noise (1500 Hz) with the tone pips (500 Hz). Kileny (1981) successfully used high pass noise with 500 and 1000 Hz tone bursts with 2 cycle rise/fall and no plateau, to estimate hearing thresholds in subjects with high frequency loss. This method has been used particularly with lower frequencies (eg 500 Hz), the assumption being that frequencies below that tested will not contribute significantly to the response. This assumption, however, may be questioned, especially at high intensities where the spread of energy to lower frequency regions is a well documented phenomenon.(Gorga et al., 1983).

d) TONES IN NOTCHED NOISE:

The use of notched (band reject) noise with the middle of the notch corresponding to the nominal frequency of the tone burst is another option for frequency specific ABR work. A commonly used method of evaluating the ability of a stimulus to elicit a frequency specific response is to measure the response latencies at different frequencies. A prolongation of wave V latency would be expected as frequency decreased and the response originated from more apical regions of the basilar membrane. Picton et al (1981) have shown that ABRs evoked using tone pips in notched noise show increased latencies in response to decreases in both frequency and intensity. The magnitude of the shifts in latency with frequency reduction correlate well with the expected travelling wave delay, suggesting that the notched noise paradigm does provide frequency specificity (Picton, 1981, Sohmer & Kinarti, 1984). The accuracy of this technique is to within about 15 dB of the behavioural thresholds in normal hearing people, though so far little work has been carried out with hearing impaired subjects (Sohmer & Kinarti, 1984). A possible problem with notched noise techniques arises through the upward spread of masking from the low frequency edge of the notch especially at high intensities (Picton et al., 1979).

e) COMPARISON OF TECHNIQUES:

Picton et al (1979) have used unfiltered white noise in the same way as notched noise. They found that response amplitudes decreased, making accurate measurement of the peaks more difficult. On the other hand, the advantage of white noise is the relative simplicity of its application when compared to other masking techniques.

In an attempt to find which form of stimulus was the most effective, Stapells et al (1984) evaluated the utility of 8 evoked potential techniques in obtaining threshold information at all audiometric frequencies (.5,1,2,4 kHz). They found that tones required less masking noise than clicks, making them clinically safer in terms of possible noise induced damage, as well as being more comfortable for the patient. When presented at repetition rates of 40/second, low frequency responses were larger and clearer making threshold identification easier, due to wave V being superimposed on the 40 Hz response.

Stapells et al (1985) suggested a compromise with notched noise being recommended for higher frequencies (1,2,4 kHz) and highpass noise used at 500 Hz. The advantages of using high pass noise at 500 Hz are that it eliminates the problem of upward spread of masking inherent for notched noise, and also requires lower overall intensities of masking than notched noise. Stapells et al point out that this lowering of overall noise intensity also reduces any possible attenuation of low frequency hearing caused by contraction of the middle ear stapedius muscle (p.170).

MASKING LEVELS:

With all masking techniques, a crucial parameter is the sound pressure level of masking noise required to effectively mask the ABR at the unwanted frequencies. A feature of the research conducted so far has been the variety of different levels of masking employed. For example, the signal-to-noise ratio used by

Picton et al (1979) and Beattie & Boyd (1985) was +25 dB, Klein (1986) used +10 dB, while Kileny (1981) used 0 dB with the noise and tone pip sound pressure levels being equal.

For the purposes of this study, and that of Purdy et al (1988), a preliminary study was carried out by Houghton (1987) to establish effective masking levels for tone pip ABRs using white noise. This work included measures of the sound pressure levels required to obliterate an ABR response elicited by tone pips of different frequencies. A resultant signal to noise ratio of -5 dB between the SPL of the tone pip and the SPL of masking noise was found to effectively mask all responses in normal hearing listeners. The present study used this signal to noise ratio.

To summarise, one of the major problems facing frequency specific ABR research is that of which stimuli will optimise both the ease of testing and the accuracy of the response elicited. Within stimulus parameters, the dilemma exists between the need for a rapid onset stimulus to provide the neural synchronicity required for a clear response - and the need for a concentration of stimulus energy to provide frequency specificity. Various masking paradigms have been introduced to allow for a rapid onset and yet still restrict responses to limited areas of the basilar membrane.

3) RECORDING FACTORS

Yet another "signal to noise ratio" problem arises when one considers how the ABR is recorded. Being a far field recording (the electrodes are placed at some distance from the activity being recorded) the ABR is susceptible to concealment in a background of electroencephalographic (EEG) and electromyographic (EMG) noise. In order to extract the relatively tiny evoked potential of interest, the relative ratios of evoked response and background noise must be optimised to allow easy and accurate identification of even very small near-threshold responses. This optimisation involves improving the signal-to-noise ratio of ABR to EEG/EMG and is achieved in part by adding and averaging a large number of stimulus presentations - usually in the order of 1000 to 4000 repetitions (Stapells et al., 1985). Because for each of these repetitions the ABR component is constant while the EEG noise is random, over successive presentations the averaged ABR signal increases at a greater rate than the noise. Averaging alone, however, does not eliminate the extraneous noise, and a number of other parameters are manipulated in order to optimise ABR measurement.

a) FILTER CHARACTERISTICS

Like all evoked responses, most of the energy in the actual ABR lies within a relatively small band of frequencies of the ongoing EEG (Jacobson & Hyde, 1985). Eliminating any energy outside of this range through the use of a band pass filter can dramatically reduce the signal-to-noise ratio before the averaging process occurs, and improve the resolution of the resultant ABR tracing (Gibson, 1978). Input filters are standardly used in all ABR work to remove as much unwanted electrical interference as possible. Schwartz and Berry (1985) reviewed studies looking at what range of frequencies contained most of the ABRs energy and reported a general agreement of between 50 to 1000 Hz. Many studies have found that ABR responses to low frequency stimuli are enhanced by lower cut-offs on the high pass filter (Stapells & Picton, 1981, Suzuki & Horiuchi, 1977). Houghton (1987) also reviewed studies looking for optimal settings of high and low pass filters. She reported that a high pass setting of between 20 - 30 Hz and a low pass setting of around 3000 Hz was optimal for frequency specific testing.

b) REPETITION RATE

Increasing the repetition rate of the stimulus has been shown to increase the latency of the response and decrease the response amplitude (Jewett & Williston, 1971). Stapells et al (1985) recommended a repetition rate of about 40/second for tone pip ABRs in notched noise. This choice of rate enables Wave V to be superimposed on the 40 Hz response and increases the overall amplitude of the response. Compared to standard ABR repetition rates of 10 to 21 clicks per second, this faster rate of presentation also reduces the time taken for each test.

c) ARTEFACT REJECTION

A safeguard against the masking of the ABR by movement artefacts or other noise is to reject any averages where the signal exceeds a certain preset voltage (Gibson, 1978).

Discarding these extreme recordings, which are commonly caused by myogenic activity, helps protect a slowly building average from being swamped in artefact.

d) STIMULUS POLARITY

The decision as to whether the stimulus is presented with a positive, negative or alternating onset phase is mostly a matter of clinical preference (Schwartz & Berry, 1985). One possible advantage of using an alternating polarity is that any electrical stimulus artefact or cochlear microphonics picked up by the recording electrodes is effectively cancelled out by adding positive and negative going stimuli (Arlinger, 1981, p.44). Davis et al (1985) reported no differences in the visual detection of responses related to the polarity of stimulus used. They did report however that, at high intensities, alternating polarity can produce stimulus artefact that becomes larger than the actual responses. This might affect the recordings of patients with more severe hearing losses, especially at or around threshold.

CONTRALATERAL MASKING

The need to prevent the crossover of high intensity stimuli to the opposite cochlea during audiometric testing at high intensities is well established. Numerous studies have demonstrated the effects of crossover in unilateral hearing losses, and also that this crossover can be obliterated by the use of contralateral masking where necessary (Ozdamar & Stein, 1981, Humes & Ochs, 1982). Humes and Ochs demonstrated also that contralateral masking has no effects on ABR latencies or amplitudes in normal hearing subjects. The need to use contralateral masking is evident whenever the signal in the test ear is louder than the non- test cochlear threshold plus the individuals interaural attenuation for the test signal.

Accurately identifying the ABR response and tracking it down to low amplitude threshold levels requires the setting of recording parameters to maximise the signal-to-noise ratio's and produce accurate and "clean" response traces.

SUMMARY:

After a comprehensive review of previous studies, Stapells et al (1985) made the following recommendations. Firstly that the most appropriate stimuli for frequency specific testing were brief tone pips - ideally having a "2-1-2" cycle rise-plateau-decay - in conjunction with notched noise. They proposed that masking noise levels be set to 20 dB below the peak equivalent SPL (peSPL) of the tones with notch widths of 1 octave and notch depths of at least 20 dB. To increase overall response size, a stimulus repetition rate of around 40/second was recommended, enabling wave V of the ABR to be superimposed on the 40 Hz response. Finally, a wide bandpass filter setting of approximately 20-2000 Hz on the recording amplifier would allow inclusion of both the ABR and 40 Hz response components.

These recommendations were followed as closely as possible by Purdy et al (1988) who set out to provide norms for frequency specific ABR thresholds in adults, for tone pips in high pass and notched noise. Due to equipment limitations, "2-1-2" tone pip envelopes were not possible at 2 kHz and 4 kHz and so 1 msec rise-plateau-fall envelopes were employed at these frequencies. The other major deviation from the Stapell's group recommendations was in the level of masking noise used, with a -5 dB signal-to-noise ratio being employed following earlier work by Houghton (1987). The results of interest to the present study are shown in Table 1 below, which gives the average masked ABR threshold for 20 normal hearing subjects at each frequency (See Table 1).

Frequency	500 Hz	1000 Hz	2000 Hz	4000 Hz
Average ABR Threshold.	36.6	30.0	28.8	28.6

TABLE 1. The average ABR thresholds (dB peSPL) to tone pips in notched and high pass noise for normal hearing (0dB HL) adults.

(from Purdy et al., 1988)

Using the same stimulus and recording parameters as the Purdy study, the rationale for this study was to investigate the relationship between pure tone audiometric thresholds and masked tone pip ABR thresholds in subjects with different degrees and configurations of hearing loss. This was done by expressing both variables in terms of their elevation from normal figures - behavioural elevation relative to 0 dB HL (re: ANSI 1969) and ABR elevation being relative to the figures in Table 1. The main question to be examined was that if normal listeners with 0 dB audiometric thresholds have a mean ABR threshold of "x" dB peSPL at a particular frequency, does someone with "y" dB sensorineural hearing loss have an ABR threshold of "x + y" dB peSPL at that frequency.

METHOD

SUBJECTS

Thirty adults (14 males , 16 females) were selected from the files of the National Audiology Centre, to comply with the criteria set down for each of the three experimental groups. The average age of the subjects was 48.6 years, and the range was from 25 to 64 years. Twenty subjects listened with their left and ten with their right ears. Subjects were selected to fit the following criteria which designated the three groups used in the investigation. Ten subjects served in each group.

a) High frequency losses - defined as having audiometric thresholds such that

i) the 2 kHz threshold was 20 dB above 1 kHz and/or

ii) 4 kHz was 40 dB above 1 kHz

b) Low frequency losses - where

i) the 1 kHz threshold was 20 dB above 2 kHz and/or

ii) 500 Hz was 40 dB above 2 kHz

c) Flat losses - where thresholds at 0.5, 1, 2, and 4 kHz were all within a 15 dB range.

Due to a difficulty in obtaining flat and low frequency subjects complying to all test criteria, the following three subjects were included:

1: A FLAT subject with a 20 dB range (40-60 dB HL) from 0.5-1 kHz

2: A LOW subject with a 15 dB drop from 2-1 kHz and a 20 dB drop from 2-0.5 kHz.

3: A LOW subject with a 15 dB drop from 2-1 kHz and a 25 dB drop from 4-1 kHz (this subject had a 15 dB increase from 1-0.5 kHz)

The average audiograms obtained from each group are shown an Figure 4. No subjects had audiometric configurations in the test ear suitable for any group other than the one they were included in, though in two cases, the non-test ear did conform to another groups criteria. Subjects were contacted by telephone and informed of the nature of the research. The NZ National Foundation for the Deaf provided payment from its Deafness Research Fund at a rate of $5 an hour. This was posted out to subjects upon completion of the study.

EQUIPMENT

Pure tone audiometry was conducted using a Grason Stadler 1704 audiometer using the right headphone of a TDH 50 earphone set for both ears. This headphone had been calibrated in accordance with ANSI S3.6 -1969. Audiograms, impedance and ABRs were obtained in test rooms with ambient noise levels complying with standard ANSI S3.1-1977 and ISO 6189-1983(E). Impedance audiometry was conducted on an Interacoustics Impedance bridge (model AZ7).

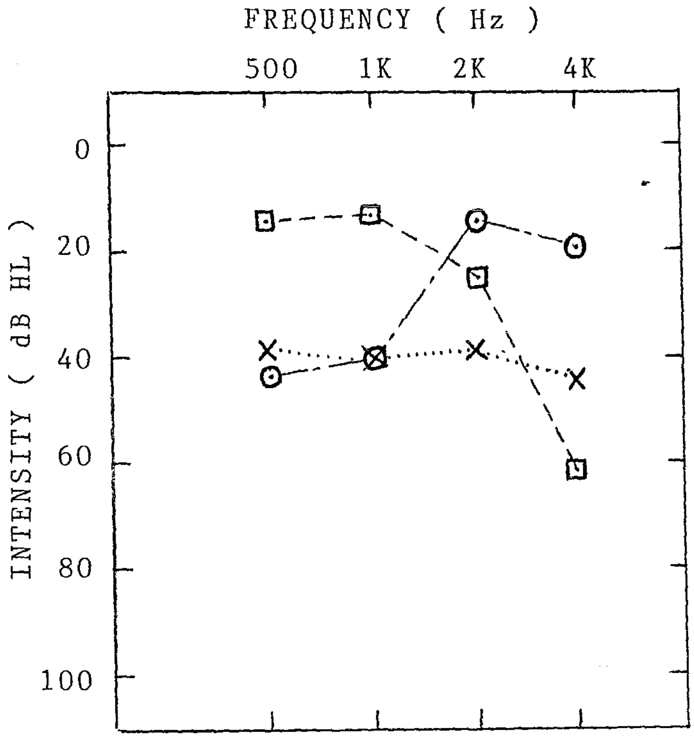


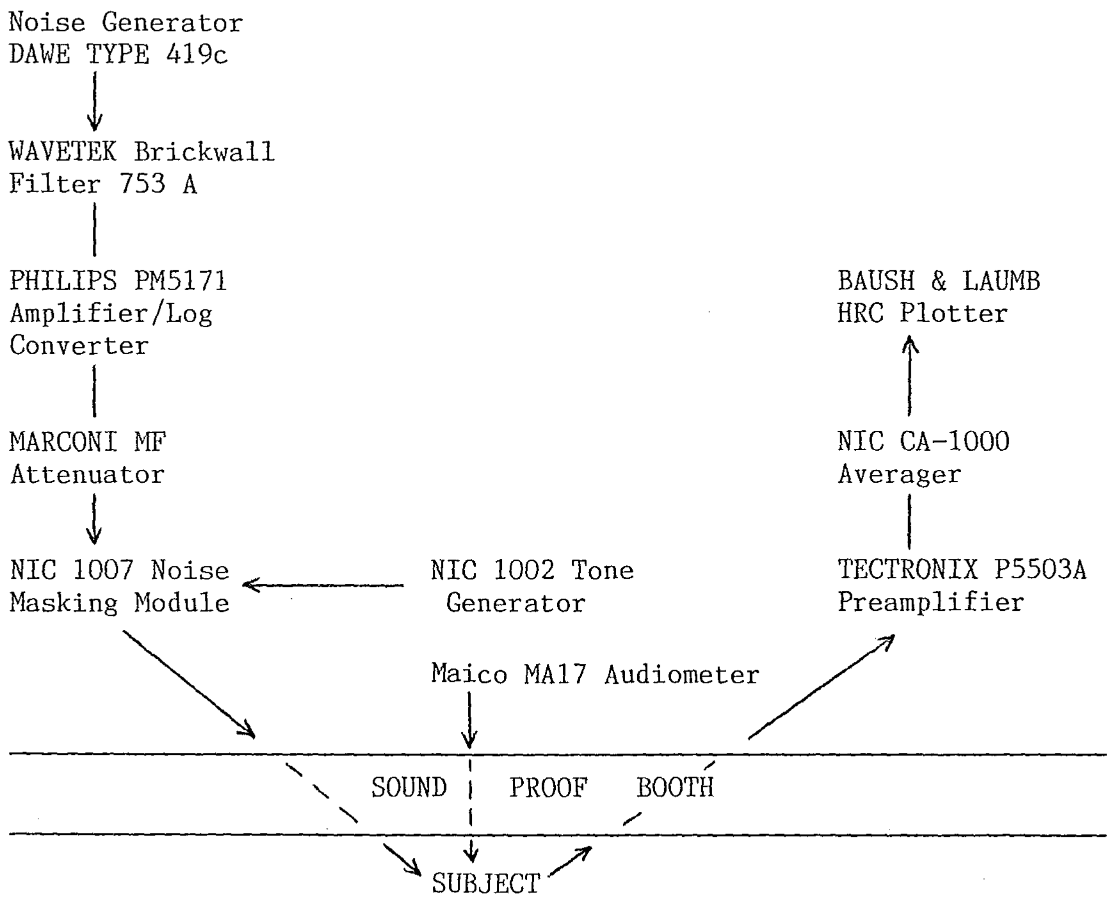
FIGURE 4. The average audiometric thresholds for each of the 3 groups tested :

⊙ - LOW FREQUENCY LOSS

⊡ - HIGH FREQUENCY LOSS

× - FLAT LOSS

FIGURE 5. Schematic layout of ABR equipment



The layout of the ABR equipment was as follows and is schematised in Figure 5.

(a) Sound Production

The tone pip signals were produced by a Nicolet NIC-1002 tone generator. Masking noise was provided by a Dawe Type 419c white noise generator (20-20,000 Hz) and passed through a Wavetek Brickwall filter (model 753 A) for high pass or notch filtering with filter slopes being 115 dB/octave. The noise was amplified by a Philips PM5171 amp/log converter ,attenuated by a Marconi MF attenuator (TF 2162), and mixed with the tone pips. Mixing was performed on a Nicolet NIC-1007A noise masking module and the output of the module delivered monaurally to unshielded TDH49 earphones mounted in MX41/AR cushions at a rate of 41.7 pips per second.

Contralateral masking was provided by a Maico MA-17 audiometer and delivered via a standard audiometric insert. It was introduced whenever the level of the tone pips exceeded 20 dB HL at the contralateral cochlea. For this calculation a conservative interaural attenuation of 40 dB for the tone pips was assumed. Appendix 1 contains details of the test ear stimulus levels which required contralateral masking, and the corresponding masking levels used.

(b) Stimulus Parameters

Thresholds at the octave frequencies of 500, 1000, 2000, and

4000 Hz were obtained in both audiometric and ABR testing. The tone pip envelope was reset at each frequency to correspond to a two cycle rise and fall and a one cycle plateau. Due to the equipment limitations mentioned earlier, envelopes at 2 and 4 kHz were set at 1-1-1 msec, while at 0.5 and 1 kHz the envelope parameters were 4-2-4 msec and 2-1-2 msec respectively. All tone pip stimuli were presented with an alternating onset phase at zero crossing of amplitude and via a linear onset ramp. Tone pips were calibrated by matching the peak to peak voltage of the tone pip plateau with that of a continuous tone of known sound pressure level. This was done using a Bruel and Kjaer 221 sound level meter linked to an oscilloscope. The resultant levels were expressed in terms of 'peak equivalent sound pressure level' (peSPL). The overall SPL of masking noise was set to be 5 dB higher than peak equivalent SPL levels of tone pips. For 500 Hz stimuli, the high pass filter was set at 750 Hz. The low and high pass filter settings at other frequencies were 590 - 1700 Hz at 1000 Hz, 1200 -3400 Hz at 2 kHz and 2300 and 6700 Hz at 4 kHz. All these settings correspond to the 3 dB down points of the filter in question.

Figure 6 shows the temporal waveforms of the tone pips used, and the spectra of the the notched and high pass noise overlayed upon the spectra of their corresponding tone pips.

(c) Recording Equipment

Electrodes were connected to a Tectronix P5503A pre-amplifier, the output of which was delivered to a Nicolet

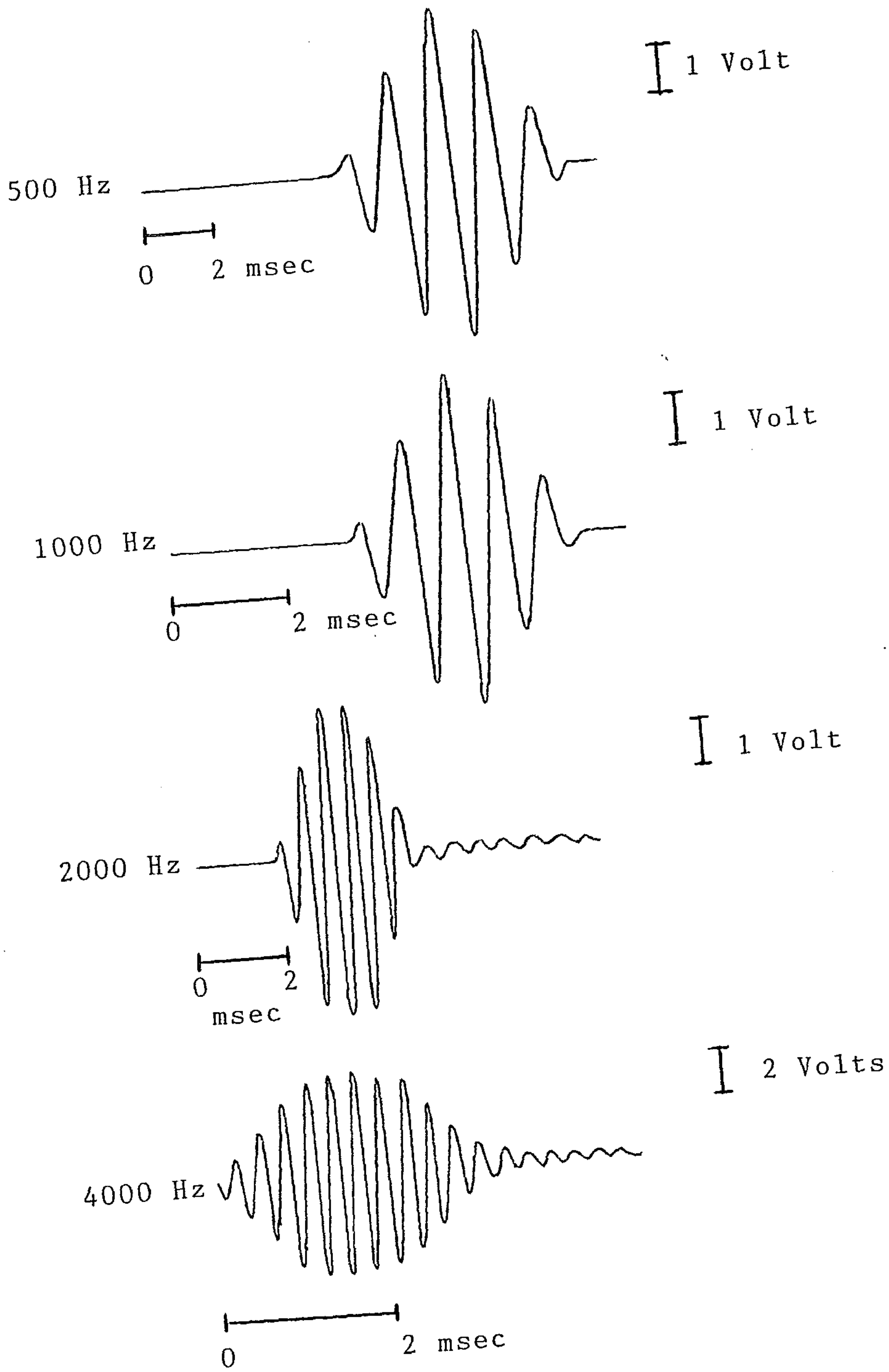


FIGURE 6 (i). Temporal waveforms of the tone pips used at each of the four frequencies.

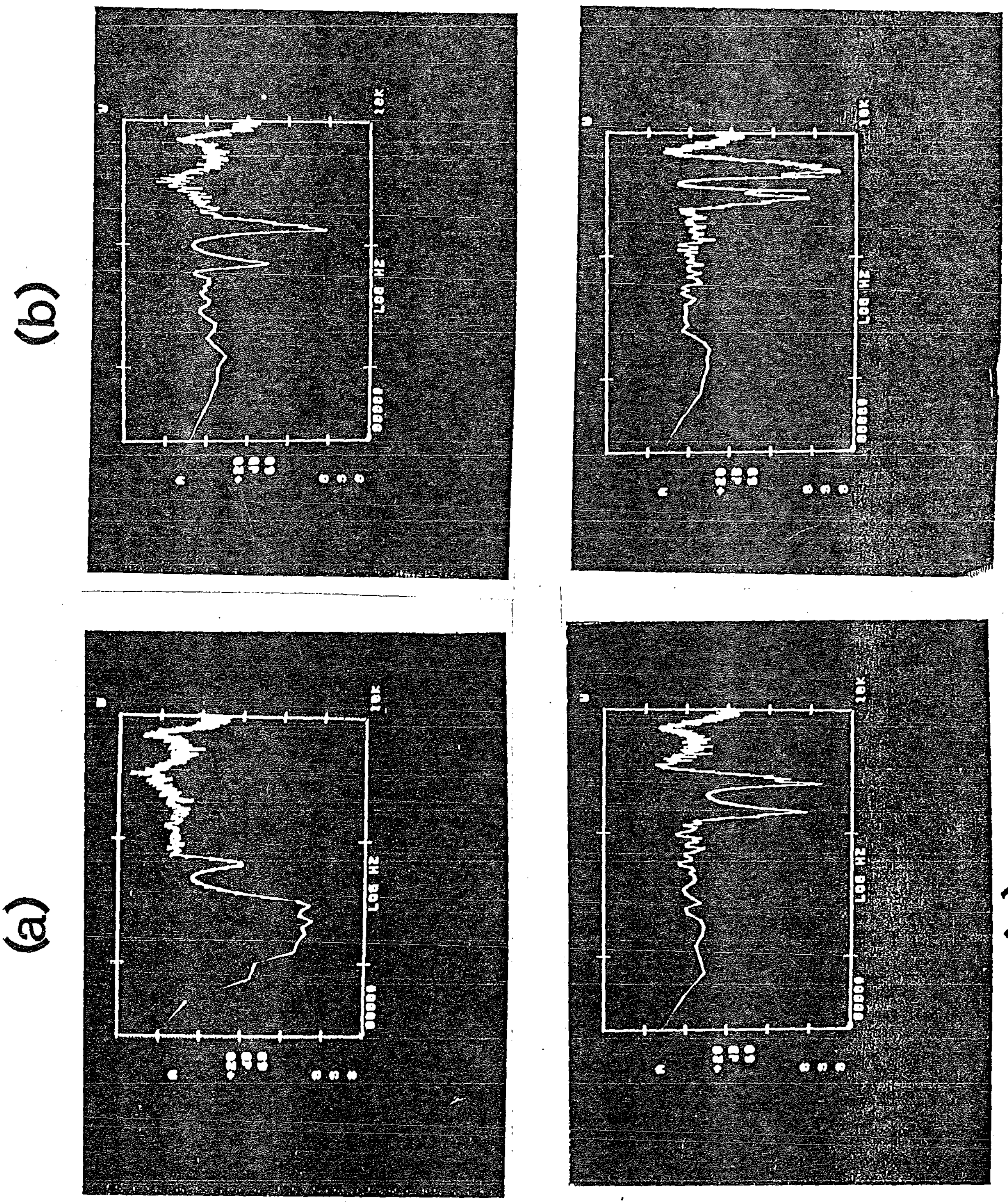


FIGURE 6 (ii). Photographs of the acoustic spectra of tone pips and masking noise.
a) 500 Hz in high pass (>750)
b) 1000 Hz in notched (590-1700)
c) 2000 Hz in notched (1200-3400)
d) 4000 Hz in notched (2300-6700)

CA-1000 signal averager. The resultant average was displayed on an oscilloscope and recorded on a Baush & Lomb X-Y plotter (Omnigraphic series HRC). Preamplifier gain was set to 10,000 with an input filter setting of 30 - 3000 Hz (12 dB/octave roll off). The input sensitivity was set at 10 or 25 microvolts full scale, depending on the noise levels in the ongoing EEG. The time epoch for each average was set at 25 msec for all frequencies except 500 Hz where a 35 msec window was used. This was done to allow for any increased latencies associated with the low frequency stimuli. A minimum of 1000 sweeps (stimulus presentations) was required in each average, and up to eight separate averages were obtained at each level (to allow an assessment of true response replicability) if response amplitude and clarity was low.

PROCEDURE

a) Audiometry: Using the same headphone for both ears, pure tone thresholds were obtained using ISO/DIS 8253.2 .(ANSI S3.21-1978) at 250, 500, 1000, 2000, 4000, 8000 Hz. If any of these thresholds were more than 5 dB different from those recorded in the previous audiogram, bone conduction was performed to establish the nature of the variation (conductive or sensorineural).

b) Impedance screening: All subjects had tympanograms performed on both ears to check for normal middle ear function as defined by:

i) Middle ear pressure between −100 and +50 mm H_2O

ii) Maximum compliance between 0.3 and 1.6cc

Of the thirty subjects tested, one failed to comply with these criteria, having middle ear pressure of −200 mmH_2O. (He belonged to the low frequency loss group and due to a shortage of subjects was allowed to remain in the study.) A further subject could not be tested due to the lack of an adequate seal for successful impedance testing. As no air-bone gap was evident in the audiogram, it was assumed that no conductive component was present.

c) ABR preparation: Subjects were seated in a sound proof booth, and the three recording electrodes were attached. During electrode application, subjects were instructed as to the procedure of the testing, asked to relax and encouraged to sleep if possible. Head or neck supports and a footrest were used, while subjects were asked to adjust the back of the chair to a comfortable position. Subjects were informed of an intercom which was linked to the equipment room and allowed constant monitoring of the subject by the tester. The booth was darkened during testing to encourage relaxation and sleep. The subject's skin was prepared by cleaning with a sterile alcohol preparation and a light abrasive (Omniprep). This ensured good electrode contact and reduced electrode impedances. Silver EEG cup electrodes were used, application via Grass EC2 electrode cream. The recording configuration for the electrodes was vertex (positive) to ipsilateral earlobe (negative) with the contralateral earlobe serving as ground. Electrode impedances were tested and balanced using an Amplaid Electrode Impedance tester to be less than 3000 Ohms to maximise recording sensitivity.

d) Click ABR: A standard unfiltered click ABR was performed and compared to previous tracings on record. If no ABR had been conducted in the past, waveform latencies were checked against established norms.(Greville, 1982). All subjects were tested at 80 or 90 dB HL with 100 microsecond rarefaction clicks at 21.7/second, input filter settings of 150-3000 Hz (12 dB/ octave rolloff), and input sensitivity of 10-25 microvolts. The response window for click ABRs was 10 msec.

e) Tone pip ABRs: ABRs to filtered tone pips in notched or high pass noise were performed on the test ear with contralateral white noise introduced where necessary. Frequency order was randomised where audiometric thresholds were below 50 dB HL. When thresholds were greater than 50 dB HL the tone pips were presented in order of best to worst (lowest to highest) thresholds. The reason for this was to minimise any disturbance of subject relaxation due to the high level of masking noise required with high thresholds. Each frequency was entered at approximately 40 dB sensation level (SL). Successively higher levels were used if necessary until a repeatable response was obtained. A descent was then performed in 5-20 dB steps with at least 2 repeatable responses being required for descent to continue. When no repeatable response was evident, runs at levels 5 dB either side of 'threshold' were made, with further descents being made if any repeatable waveforms were produced. The latencies of all repeatable peaks were labelled on the tracings, and baseline runs with no stimuli were performed regularly to provide judges with an estimate of signal to noise ratio's present during each individual session.

Depending on the subjects ongoing EEG and general response state, input sensitivity was varied from 10-25 microvolts to maximise the signal to noise ratio. If myogenic activity was excessive, testing was stopped and where necessary breaks were taken to alleviate any subject discomfort or restlessness. When subjects slept, testing continued for as long as possible. Up to 3 subjects were seen daily, with the entire test procedure lasting on an average of 3 hours per subject. Sleeping subjects were generally finished earlier than this, while restless subjects often required more time. Upon completion of testing, electrode balances were rechecked before removal. Subjects were thanked and sent home.

f) Judging : All recordings were plotted and presented to 3 judges experienced in assessing clinical and experimental ABR tracings. The tracings were labelled so that the intensity of the stimulus was not visible, and were ordered from 1-9 with 1 being the most intense and 9 the least intense level. Judges were required to nominate the last trace at which a response was present. This definition of threshold has been used by previous studies conducted at this laboratory (Houghton, 1987, Purdy et al., 1988) and is a standard used in most clinical situations. Judges were given the following guide lines based on those used by Houghton and Purdy outlining important aspects of brainstem evoked potential responses to tone pips.

1. Waves must be clearly visible and replicable.

2. Wave V is identified as the maximum positive peak occuring between 6 and 18msecs after tone onset.

3. The maximum positive peak is followed by a negative trough. In the case of 500 Hz tonepips, this negative trough is often the most prominent feature.

4. Response amplitude decreases as intensity decreases.

5. Response latency increases as intensity and frequency decreases.

Judges were also informed that the stepsize between any two tracings could range from 5 - 20 dB. Upon request, the size of any given step was available to each judge, though absolute intensity was not revealed. Latency and frequency information were available on all traces while the order in which subjects were judged was randomised.

All sets of traces were judged twice by each of the judges and a total of 240 threshold decisions thus obtained. After the initial sets of tracings were judged, subjects where total disagreement or large partial disagreement occurred were asked to come in for retesting at the frequencies where the judges could not agree. Unfortunately due to the time restraints imposed on the researcher, subjects could only be retested once, and so the second set of traces was used in the final data calculations. Retesting involved a repeat audiogram as specified above, a click ABR, and tone-pip ABR recordings at the required frequencies. In all, 9 subjects required some degree of retesting, 3 coming from each experimental group.

RESULTS

RESPONSE MORPHOLOGY

Figure 7 (a-e) presents a sample series of traces obtained from a high frequency group subject, (FB) for each of the four frequencies. The morphology of the responses at 500 Hz and 1000 Hz demonstrate the superimposition of the ABR on the 40 Hz steady state response. At 2000 and 4000 Hz the response is dominated more by wave V of the ABR, which is consistent with previous masked ABR work (Stapells et al, 1984). At all frequencies, decreases in intensity are associated with diminishing definition of the vertex positive peak and also increasing peak latencies. A striking feature of high intensity runs was the presence of stimulus artefact on the tracings, an example of which can be seen in Figure 7 (e). This artefact decreased the clarity of the response at high intensities though in most cases it did not affect threshold judgement, as lower intensity responses were always obtainable without the artefact present. This problem was reported also by Davis et al (1985) when using an alternating onset stimulus polarity at high intensities.

In all, 120 threshold decisions (30 subjects x 4 frequencies) were made by each of the 3 judges during the initial assessment. On a single occasion, one of the judges did not identify any repeatable response qualifying as a threshold. As both the other judges did indicate a threshold in the traces, for statistical

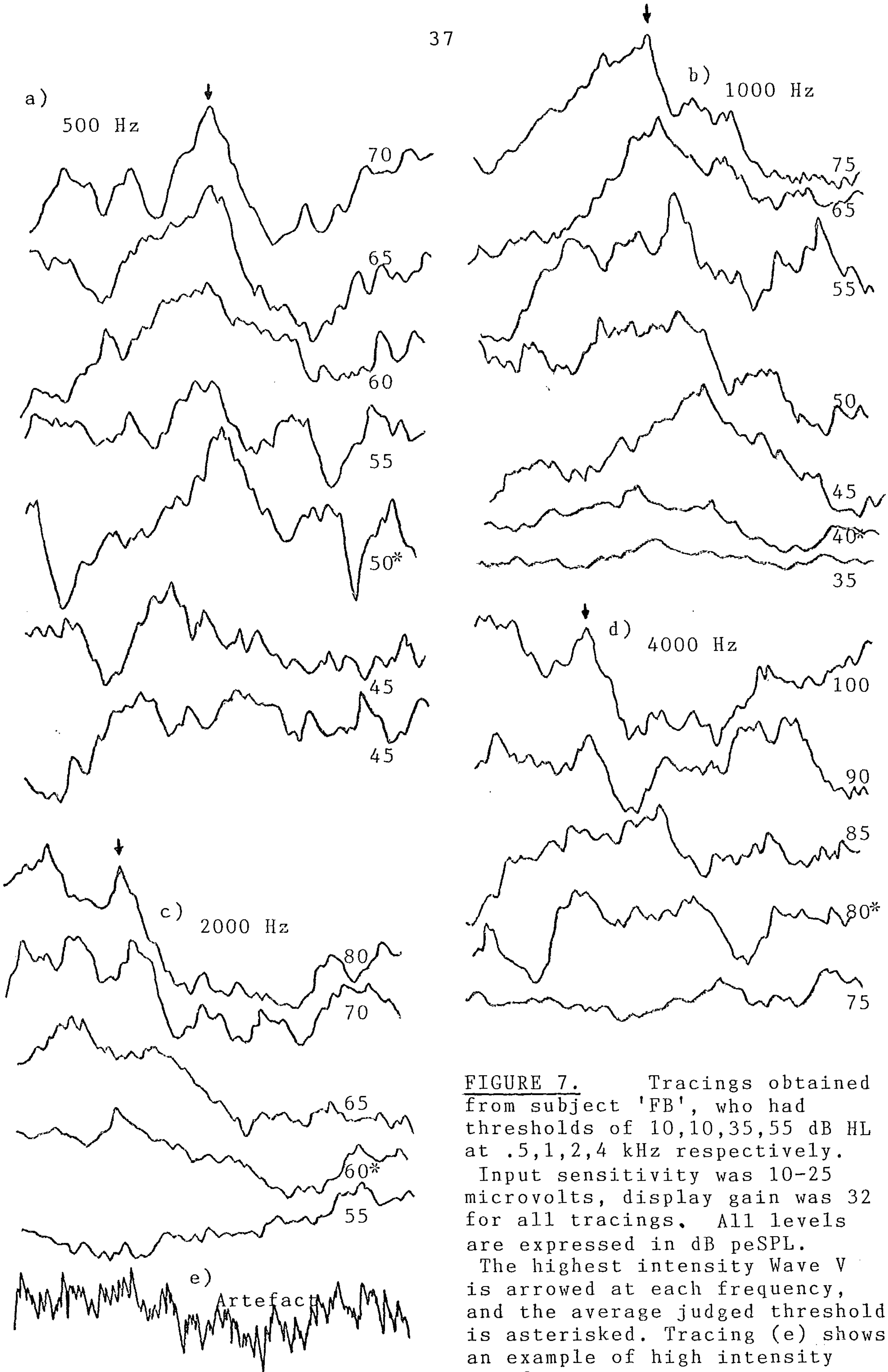


FIGURE 7. Tracings obtained from subject 'FB', who had thresholds of 10,10,35,55 dB HL at .5,1,2,4 kHz respectively. Input sensitivity was 10-25 microvolts, display gain was 32 for all tracings. All levels are expressed in dB peSPL. The highest intensity Wave V is arrowed at each frequency, and the average judged threshold is asterisked. Tracing (e) shows an example of high intensity artefact.

purposes the third judge was reckoned to have nominated a level 5 dB above the highest intensity tested (85 dB dial). Figure 8 shows a breakdown of the inter-judge performance indicating the range of disagreement between the three judges and the relative percentages. Total agreement, defined as all judges nominating the same level as threshold, was obtained on 55 occasions (45.8%). Total disagreement, defined as all judges nominating different threshold levels, occured on 8 occasions (6.6%). Where total or partial disagreement did occur, it ranged from 5 to 25 dB in size.

Because of the relatively low level of inter-judge agreement, a second set of threshold decisions was obtained from each judge to allow some assessment of intra-judge reliability. Reassessment was conducted under the same guidelines as initial assessment, and Figure 8 also shows the inter-judge breakdown of this second evaluation. Total inter-judge agreement dropped to 43 thresholds (35.8%) while total disagreement increased to 12 thresholds (9.9%). All judges were within 10 dB of each other for 97.4% and 94.1% of the thresholds in each respective set, with a combined level of 95.8% overall inter-judge agreement at the standard clinical error levelof ±5 dB.(Katz 1985)

Figure 9 shows the individual performance of each judge over the two seperate assessments. Judge 1 made identical threshold decisions on 78 traces (65%), judge 2 on 79 traces (65.8%), and judge 3 was the most consistent with 98 trace agreements (81.7%).

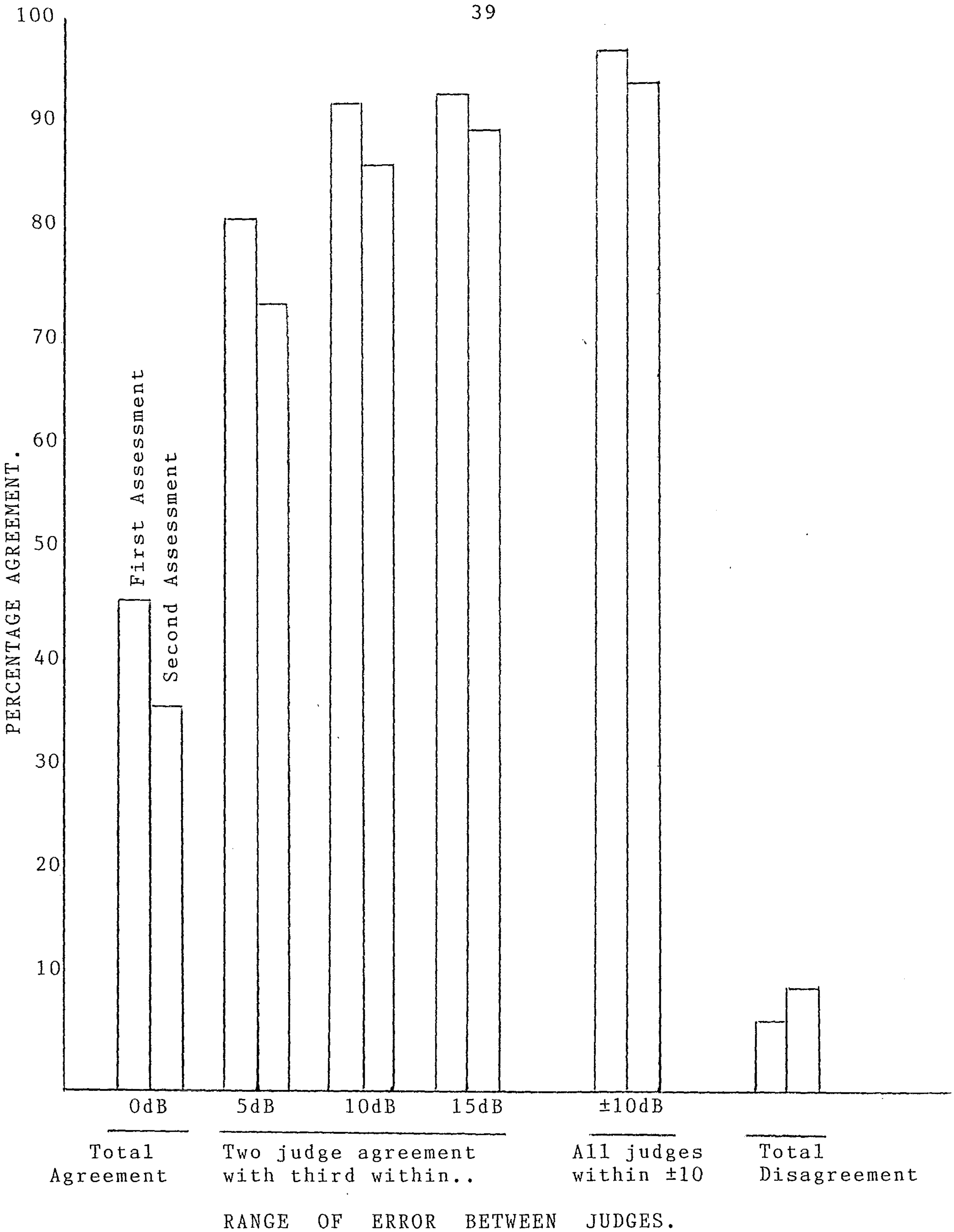
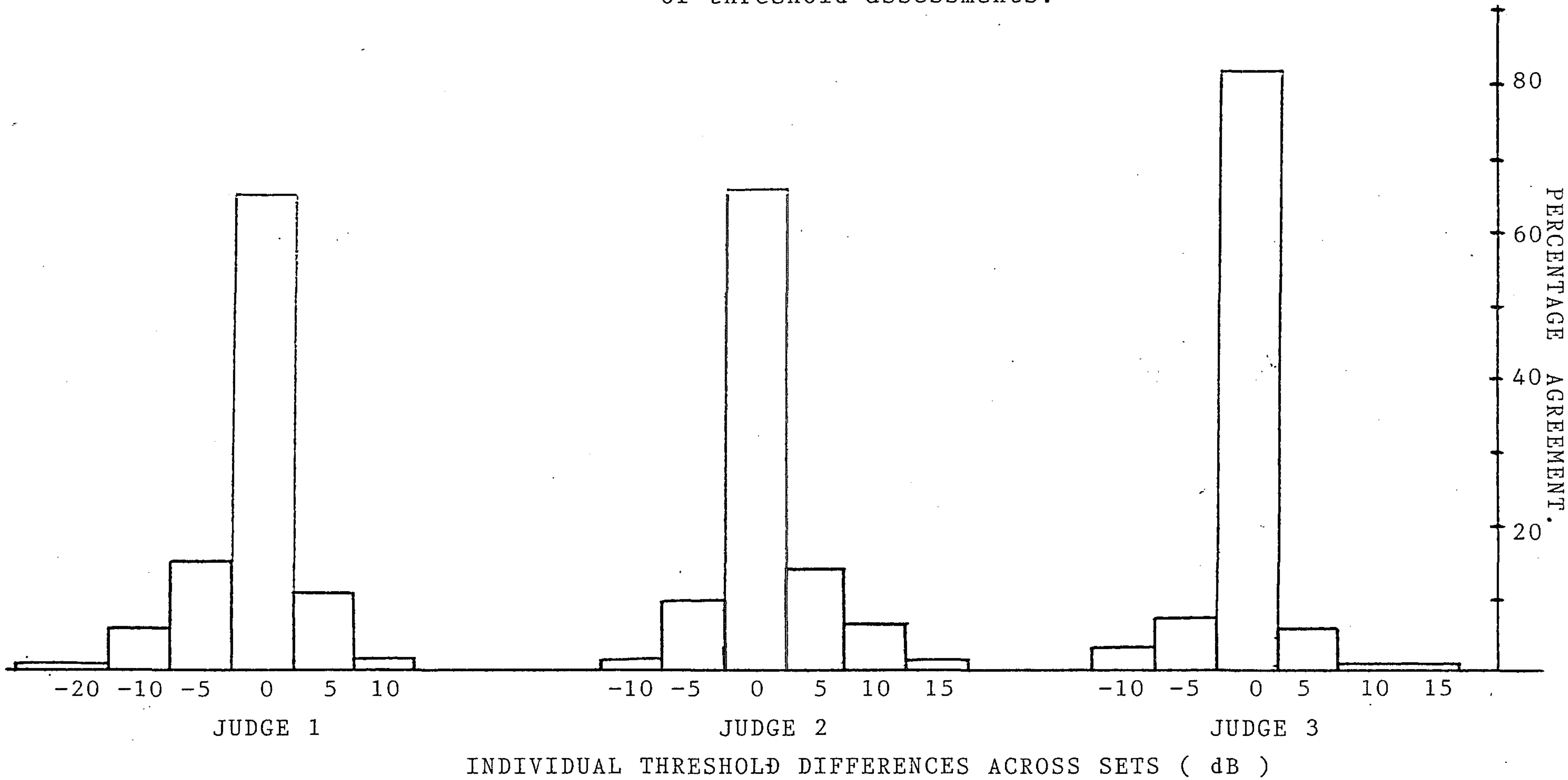


FIGURE 8. Inter-judge agreement over both assessments.

FIGURE 9. Histograms of the Intra-judge reliabilty across the two sets of threshold assessments.



Measured at the ±5 dB clinical error level, intra-judge agreement for the three judges was 89.1%, 89.9%, and 95% respectively, indicating relatively high intra-judge agreement. Because of the inherent variability in threshold assessment both within and across judges however, it was decided that an average of all six measures at each level (2 sets x 3 judges) would be used as the 'judged physiological threshold'.

Two further points regarding the ABR response traces should be noted. Firstly, the quality and clarity of response traces exhibited great variation both across and within subjects. This was most evident when sleeping subjects awoke and smooth "quiet" tracings were replaced with erratic "noisy" ones. Figure 10 (a) shows two tracings obtained at the same stimulus level and with identical recording parameters for one patient near threshold. It demonstrates how easily any small responses can be "hidden" by excess EEG noise. Secondly, intermittent 50 Hz mains interference was detectable during some recording sessions. Figure 10 (b) shows an example of the effect of the interference on a recording (i), and also the interference alone (ii). Unfortunately, the source of this interference could not be traced, and though testing was halted whenever its presence was detected, the exact effects of this interference at other times was unknown.

DATA TRANSFORMATION

The judged physiological threshold was converted from dial

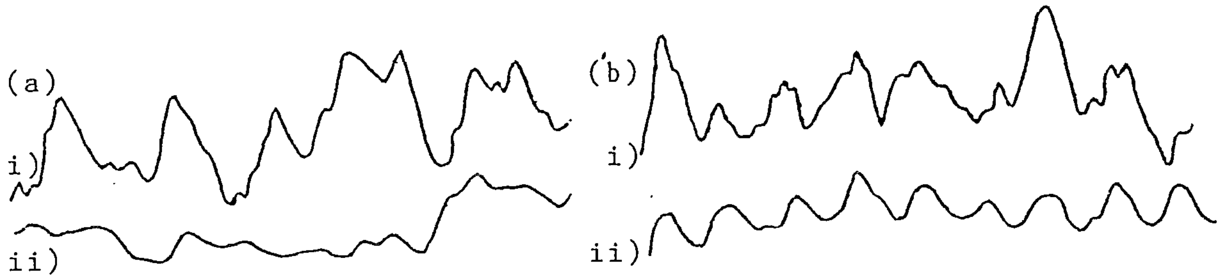


FIGURE 10. Examples of a) a 'noisy' and 'quiet' tracing recorded at the same level.
b) intermittent 50Hz mains interference i) with a tracing, and ii) alone.

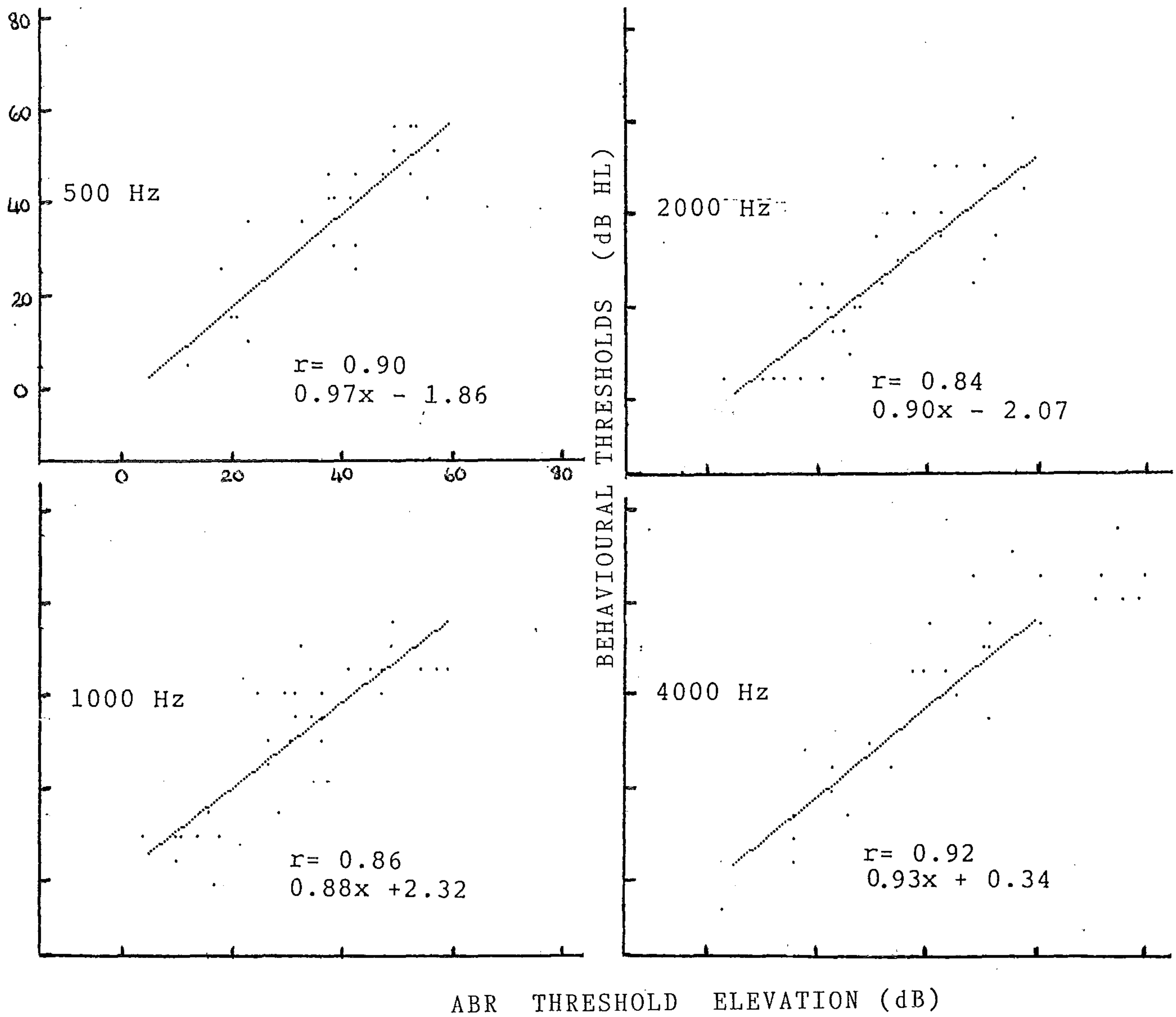


FIGURE 11. Scatterplots, correlations, & regression equations for ABR threshold elevation Vs behavioural thresholds over all four frequencies tested.

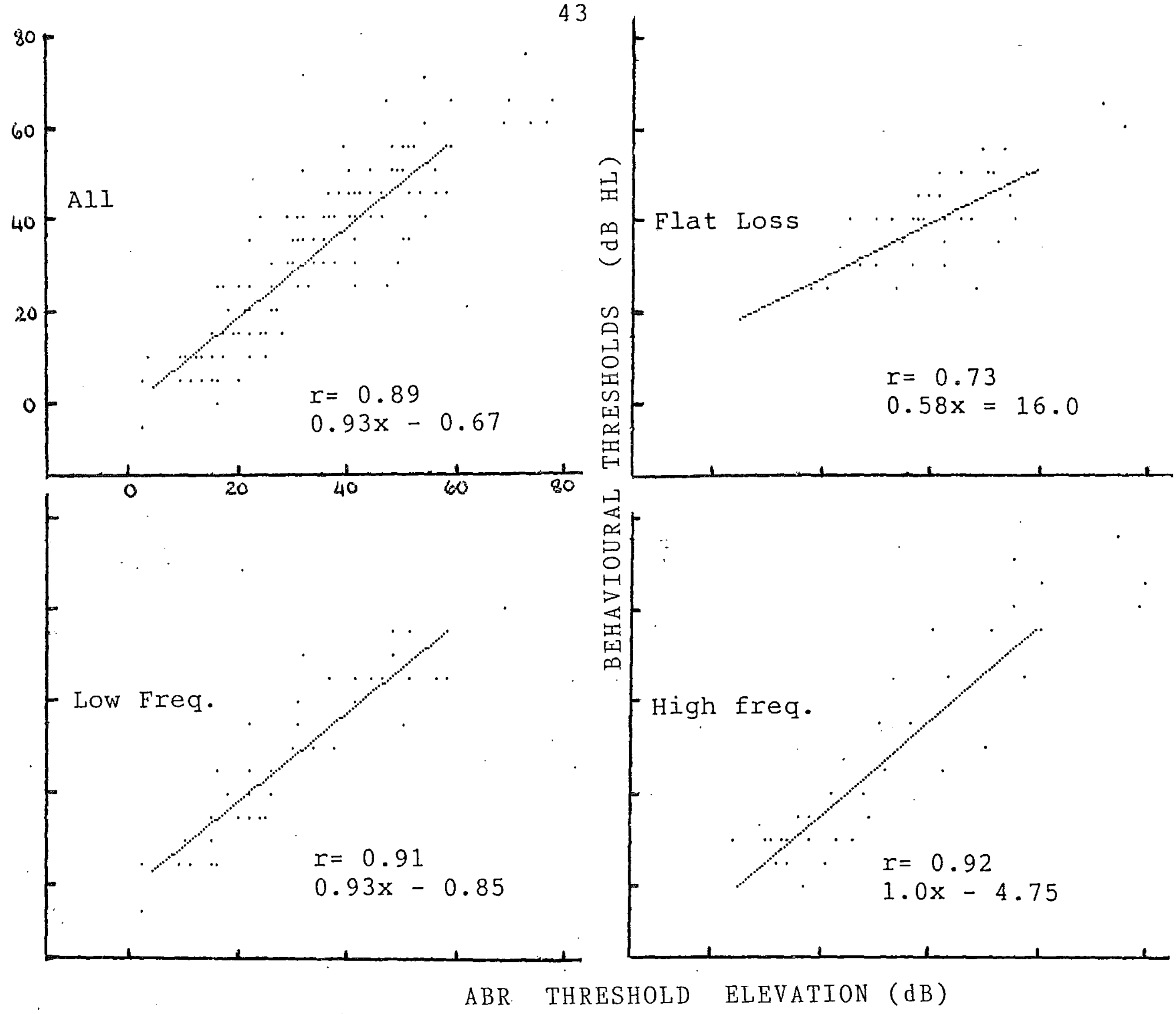


FIGURE 12. Scatterplots, correlations & regression equations for ABR threshold elevation Vs behavioural thresholds for each group and for all data combined.

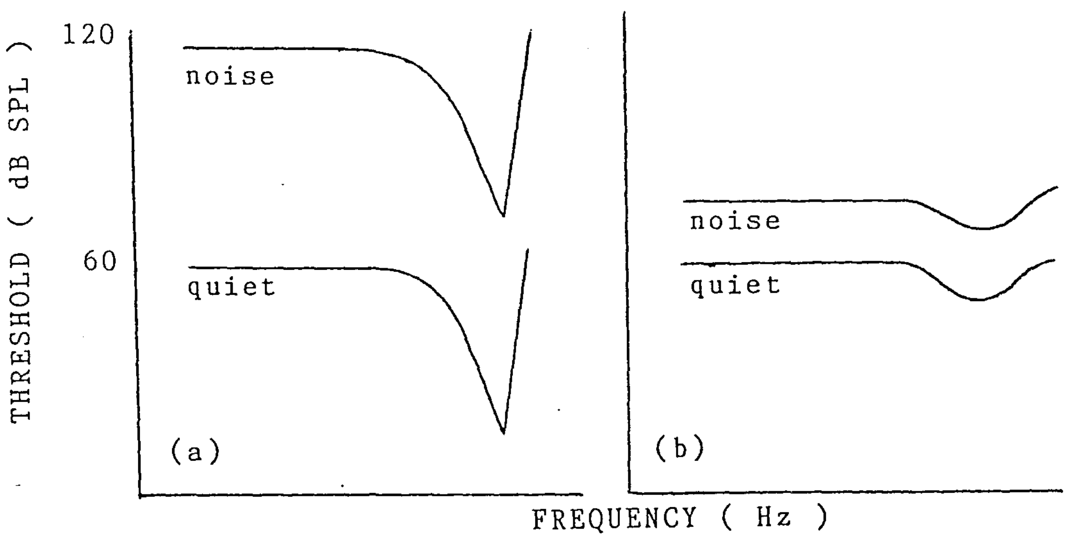


Figure 13. Shifts in tuning curves associated with a) a normal, and b) a damaged cochlea. Note that the addition of noise shifts low frequency thresholds less in the abnormal ear.
(from Gorga et al., 1983, p.355)

reading to dB peSPL by adding a constant of 30 dB at each frequency. This was based on preliminary calibration work with the ABR equipment which showed that at all frequencies, dial reading was within 30 dB (±1 dB) of the corresponding dB peSPL level. (Houghton, 1987). The resultant data expressed the average "absolute" ABR threshold in dB peSPL. These absolute thresholds were then transformed via the normative data obtained by Purdy et al (1988, See Table 1) to express the thresholds relative to the average ABR threshold for normal hearing listeners. By subtracting the correction factors in Table 1 from the absolute (dB peSPL) thresholds a measure of physiological threshold elevation (PTE) for the hearing impaired subjects was obtained. Table 2 (a) presents the mean PTE levels for each group and frequency. For final data analysis all data was rounded off to the nearest decibel integer. The relationship between PTE and behavioural thresholds was investigated by looking at the differences between PTE and behavioural thresholds and by calculating correlations and regression equations for PTE and behavioural thresholds.

Table 2 (b) shows the cell means, standard deviations and ranges for the differences between physiological threshold elevation and behavioural threshold across the 3 impairment groups and 4 frequencies. A two way analysis of variance, with repeated measures on frequencies, showed no significant effect of either impairment group ($F(2,27) = 0.999$, n.s.), or frequency ($F(3,81) = 1.252$, n.s.), nor group x frequency ($F(6,81) = 2.1966$,

a)	500 Hz	1000 Hz	2000 Hz	4000 Hz	ALL
LOW	41.8 (11.7)	40.9 (11.9)	18.6 (9.1)	25.8 (20.3)	31.8 (18.4) 2.9 to 70.6
FLAT	41.4 (12.1)	38.5 (10.1)	41.9 (9.4)	48.6 (15.4)	42.6 (12.2) 18.4 to 76.4
HIGH	20.3 (8.8)	16.5 (9.7)	34.0 (15.6)	59.9 (14.0)	32.7 (19.7) 4.2 to 79.7
ALL	34.5 (14.7) 11.7 to 57.6	31.9 (15.2) 4.2 to 60.0	31.5 (15.0) 2.9 to 57.9	44.8 (21.7) 3.1 to 79.7	35.7 (17.6) 2.9 to 79.7

b)	500 Hz	1000 Hz	2000 Hz	4000 Hz	ALL
LOW	-0.2 (6.8)	1.0 (9.3)	4.3 (6.2)	6.7 (6.0)	2.95 (7.4) -17 to 17
FLAT	2.4 (7.4)	-1.4 (8.0)	2.9 (10.7)	3.5 (6.8)	1.85 (8.2) -15 to 24
HIGH	7.1 (5.6)	4.6 (6.8)	8.4 (8.5)	-1.2 (11.9)	4.73 (9.0) -16 to 21
ALL	3.1 (7.1) -12 to 18	1.4 (8.2) -17 to 17	5.2 (8.7) -8 to 24	3.0 (9.0) -16 to 19	3.18 (8.3) -17 to 24

TABLE 2. A breakdown of cell means, standard deviations (in brackets), and also ranges for ;

a) physiological threshold elevation (PTE) relative to Purdy et al. (1988)

b) the difference between PTE and behavioural threshold elevation (re : 0 dB HL).

All data is expressed in decibels (dB).

n.s.). The lack of any group or frequency effects on the differences between the behavioural and physiological threshold elevation allow us to look at the data as a whole. Over all groups and frequencies, the average difference between behavioural and PTE elevation was 3 dB (s.d = 8 dB), though the differences spanned a wide (41 dB) range from -17 to 24 dB.

Correlation coefficients and linear equations regressing the ABR threshold on the behavioral one were calculated for each group and each frequency as well as over all groups and frequencies. These results and the individual data points are shown in Figures 11 & 12 . In addition to this simple linear regression, an analysis of the combined data was conducted to check for the presence of a quadratic trend. This analysis revealed a significant quadratic component ($t(117) = -2.781$, $p<.05$), but the addition of this component produced an increase of only 0.70 % in variance accounted for. Hence, for practical clinical purposes as is clear from the figures, the relationship between ABR and behavioural thresholds is linear.

From Figures 11 & 12, it can be seen that high correlation coefficients were obtained in all cases, ranging from 0.73 to 0.92. A high degree of similarity between regression line slopes was evident across the four frequencies, (0.88 to 0.96) while across groups the slope for the flat losses (0.58) was less than both low and high frequency losses (0.93 and 1.0 respectively). The standard error of estimate for the regression lines ranged from 6.6 dB (flat losses) to 9.1 dB (high frequency losses) with an overall error of estimate of 8.2 dB. These statistics suggest that all regression lines provided a similar if rather variable fit for the data points obtained at each frequency and for each impairment group. Incorporating data from all groups and frequencies, the best fitting linear regression line took the form

$$B = ABR \times 0.93 - 0.67$$

where B = predicted behavioural threshold
and ABR = the elevation of ABR threshold compared to normal listeners at each frequency.

Obtaining the "ABR" variable involves establishing a subjects peSPL ABR threshold, and correcting this with the appropriate frequency specific constant based on normative data from normal hearing subjects (Purdy et al., 1988). As with most electrophysiological norms, this normative data should ideally be obtained for each individual clinical setting.

In addition to the analysis of PTE, an analysis of variance was also performed using the absolute (i.e. uncorrected) peSPL ABR threshold data. Results once again showed no significant impairment group effect ($F(2,27) = 0.999$, n.s.) or group x frequency effect ($F(6,81) = 2.9$, n.s.) but a significant frequency effect was present ($F(3,81) = 8.12$, $p < 0.01$). Further analysis using Tukeys standardised range ($p=0.05$, $MS=58.22$)

test showed that the mean 500 Hz threshold was significantly higher than thresholds at all other frequencies, with no significant difference evident between other pairs of means. (See Appendix 2 for specific comparisons) This significant difference suggests that some form of freqeuncy specific adjustment is necessary in order to accurately compare ABR threshold elevations with behavioural hearing thresholds.

DISCUSSION.

The results of this study indicate that a strong positive relationship does exist between behavioural audiometric thresholds and electrophysiological thresholds measured via ABRs to tone bursts in notched and high pass noise. Correlation coefficients expressing the relationship between elevation of behavioural and physiological thresholds were consistently high ranging from 0.74 to 0.92 with an overall average of 0.89 across all groups and frequencies. These results are in keeping with previous studies investigating this relationship, though these have predominantly worked with unmasked stimuli. McGee and Clemis (1980) using "1-0-1" msec tones at 1, 2 and 4 kHz to derive audiograms in normal, conductive, and cochlear losses, found high correlations between audiometric (dB HL) and ABR (dB peSPL) thresholds. The cochlear loss subjects in their study produced linear regression lines with slopes between 0.65 and 0.80. Hayes and Jerger (1982) compared "2.5-0-2.5" msec 500 Hz and 2 kHz tone burst ABR thresholds to audiometric thresholds in hearing impaired adults and children, and found that while correlations were good, 500 Hz thresholds showed less correlation (0.71 vs 0.81) than 2000 Hz thresholds. Our data did not show any decrease in correlation with increased stimulus frequency, indeed the correlation obtained at 2000 Hz (0.84) was less than that at 500 (0.90), and the lowest correlation of all frequencies. More recently Kileny and Magnathan (1987) using

2-0-2 cycle 500 Hz tone pips with hearing impaired children, reported regression line slopes of 0.966 in relation to 250 and 500 Hz behavioural thresholds. Though direct comparisons with these studies is difficult due to the variety of different stimulus and recording parameters used, the general trends of the data appear to be the same.

The question asked at the beginning of this study was that, if a subject's ABR threshold at a given frequency was elevated 'x' dB in relation to the average "normal" threshold at that frequency, did that subject have a hearing loss of 'x' dB relative to his normal hearing peers. The overall linear regression equation obtained from this study (B = ABR x 0.93 - 0.67) expresses the best fit line for the data and on the surface, because the slope is close to one, appears to provide some support for a one to one relationship between pure tone and ABR threshold elevation. However, examination of the standard deviations and ranges of the data, as well as the standard errors of the regression lines gives a clearer idea of the true variability of the data. Data pooled over all groups and frequencies in this study showed a standard deviation of 8.3 dB and a range of 41 dB. Comparison of these data with those of Purdy et al (1988) shows that a major difference exists between the studies. In Purdy's study, standard deviations of between 3.4 and 5.0 dB were evident around the mean individual ABR thresholds in normal hearing adults, while the range was only 21 dB. Hence, there was much more variability in the differences between behavioural and physiological thresholds in the hearing impaired subjects of the present study than in the normal hearing subjects of Purdy et al (1988).

The standard error of the estimate, representing the standard deviation of the distribution of actual behavioural results around values predicted by the regression equation, was 8.22 dB for this study. This is less than the standard errors of 12.2 dB and 10.6 dB, reported by Hayes & Jerger (1982), at 500 and 2000 Hz respectively. It is interesting to note that, when looking at the differences between absolute ABR and behavioural thresholds in the present study, the difference over all groups at 500 Hz was significantly greater than at other frequencies. Similar results were obtained by both Hayes & Jerger (1982), who reported greater predictive error at 500 Hz (see above), and Purdy et al (1988) who found that ABR thresholds at 500 Hz were 5-6 dB higher than those at 1,2 and 4 kHz in normal hearing listeners. The finding of both a frequency effect on absolute ABR / behavioural threshold differences, and of less predictive error in results allowing for these differences, supports the present study's decision to use a frequency specific correction in the conversion of absolute ABR threshold to ABR threshold elevation relative to normal listeners.

Looking at the increased variability evident in our results a number of possible causes present themselves:

1) <u>JUDGE RELIABILITY AND RESPONSE CLARITY</u>

The overall inter-judge agreement obtained in this study was lower than that obtained by Purdy et al (1988) who investigated the ABR thresholds of normal hearing subjects. Purdy, using a similar judging technique obtained 68% total agreement between judges with a further 26% of the decisions within 5 dB, giving a total of 94% agreement using the standard clinical error of ±5 dB. For the present study, using the same ±5 dB tolerance, agreement ranged between 73.3% and 81.6%, however total agreement (i.e. all judges equal) dropped as low as 35.8%. Some studies using similar judging methods have also reported higher levels of judge agreement. Houghton (1987) reported 90% agreement between three judges in determining where tone pip ABR's were effectively masked by white noise. Hayes and Jerger (1982) report a 3% false positive rate for two judges seeking tone pip ABR thresholds, but do not mention actual inter-judge agreement levels. Other studies do not give any information as to how threshold judgements were made or what degree of agreement occured (eg. McGee & Clemis, 1980).

As thresholds are approached and response amplitudes decrease, the task of making threshold decisions increases in both difficulty and importance if accurate predictions are to be made clinically. Examination of the intra-judge data from this study shows that while 90% agreement was obtained within the clinical range of ±5 dB, the total agreement within two judges dropped to 65% for the same set of tracings. All judges commented on the high levels of noise often present, and it was clear that inter-judge agreement was highest for tracings obtained from sleeping subjects.

One way possibly to improve inter-judge reliability might be to standardise the response state of subjects through sedation which would reduce signal (ABR response) to noise (EEG/EMG) ratios and improve the clarity of tracings especially around threshold. However, searching for thresholds in subjects with hearing losses invariably requires much louder stimulus levels than with normals. The high levels of masking required with these stimuli often prevents sleep and precludes total relaxation especially for subjects with sloping losses whose hearing may be normal at some frequencies. A question that must also be asked is that even with high agreement as to the last repeatable response, would a lower level threshold have been obtained if response clarity at low intensities had been improved? Thus, because of unreliable response states, absolute ABR thresholds may have been masked by EEG noise, preventing any replicable peaks being visible in the tracings. Routine sedation, however, is not often a practical alternative clinically and so acheiving and maintaining an acceptably "quiet" response state in the standard testing situation remains a problem.

A second alternative for improving the consistency and reliability of threshold estimation may lie in more objective judging techniques. A series of papers has recently outlined a statistical approach for the quantitative identification of ABR thresholds (Elberling & Don, 1984, Don et al 1984, Elberling & Wahlgreen 1985). Although requiring the use of more complex equipment, methods such as this enable automatic threshold detection and can reduce the variability of test interpretation as well as maximising ABR efficiency through optimising parameters such as the number of stimulus presentations needed for response identification. (Don et al, 1984). Poor signal to noise ratios, small response amplitudes and variations in background noise levels between traces are problems inherent in all types of ABR work. Reducing the amplitude and clarity of the ABR even further by embedding the stimulus in masking noise, increases the need for more objective response detection techniques for accurate identification of physiological thresholds.

2) VALIDITY OF NORMAL Vs HEARING IMPAIRED COMPARISONS

The present study based its ABR threshold elevation figures on normative data obtained from normally hearing listeners. A number of factors challenge the validity of both using such normative data for hearing impaired populations, and also the direct comparison of electrophysiological and behavioural measures. Specifically these factors are concerned with pathological changes often associated with cochlear losses including changes in frequency selectivity and temporal integration.

Frequency Selectivity:

Briefly, the decreased "sharpness" of auditory nerve fibre tuning curves (see Figure 13) present in many cochlear pathologies, can result in reduced frequency selectivity. By this we mean a reduction in the ability of a nerve fibre to respond specifically to one particular frequency at lower intensities than other frequencies. Loss of the sharp tip evident on normal turning curves means that a given fibre's threshold to its characteristic frequency is brought back much closer to its threshold for surrounding frequencies. At higher intensities and with their increased frequency spectra, the brief tone bursts used in ABR audiometry will activate more extensive areas of nerve fibres, and therefore decrease frequency selectivity. The masking employed in this study is designed to restrict responses to specific areas of the basilar membrane and limit this energy spread. However, while there is evidence to support the various masking paradigms with normal hearing subjects, the validity of generalising these assumptions to subjects with cochlear losses has been questioned. Gorga et al (1983) point out that elevations in hearing thresholds associated with the application of masking noise to normal ears may be substantially decreased when the same amount of masking is applied to an ear with abnormal tuning curves. Figure 13 shows that a 60 dB shift in threshold in response to masking for a normally tuned fibre may only correspond to a 10 dB shift for an abnormally tuned fibre, and that the thresholds of this fibre to lower frequency stimuli are not altered to the same extent.

Hence, the effective masking levels in normal ears might not be the effective masking levels in pathological ears. Inaccurate predictions of thresholds might thus result as responses originate from areas other than that apparently under investigation. Validation of masking levels in hearing impaired subjects is required to ensure that these levels are indeed effective.

As mentioned earlier, because masked ABR tests are place specific, it is possible that they may provide a more accurate assessment of cochlear function than conventional pure tone audiometry. However, with the aim of such tests being to estimate the subjects perceptual hearing status, predictions based on place specific testing may overestimate actual hearing loss. Hence, using the example of a subject with no hair cell function beyond the basal turn of the cochlea, due to the long low frequency tail of high frequency auditory nerve fibres, the subject may be able to "hear" low frequency sounds at the level of a moderate hearing loss. The dangers of using place specific test results in this instance are that overestimating actual hearing loss can lead to the fitting of inappropriately powerful hearing aids which in turn may damage a subject's hearing further.

Temporal Integration:

It is widely accepted that people with cochlear damage encode brief stimuli differently than normal hearing people. Many studies have documented that in cochlear hearing loss - regardless of pathology - the threshold improvement associated with longer stimulus duration (temporal integration) is often reduced at threshold when compared to normals (McGee & Clemis, 1980, Chung & Smith, 1980, Gorga et al., 1984). Temporal integration has been shown to vary in size as frequency changes, and also to vary according to the degree of cochlear loss (Pederson & Saloman, 1977, Gengel & Watson, 1971).

A possible effect of the reduced temporal integration present in many cochlear losses would be that differences between long tone thresholds (behavioural) and short tone thresholds (ABR) are less than in normals (McGee & Clemis, 1980). Gorga et al (1984) have shown that while ABR thresholds are independent of stimulus duration for both normal and hearing impaired listeners, the improvement in behavioural thresholds associated with longer duration stimuli is much more marked in normals. They suggest that the two types of responses (behavioural & electrophysiological) result from "different aspects of underlying neurophysiological processes" (Gorga et al., 1984, pg 618). In using the normative data obtained by Purdy et al (1988) based on peSPL ABR thresholds inormals, the present study avoided the problems associated with the use of a behavioural reference for ABR measurements - a practice which has been criticised in the past (Gorga et al., 1984). However Pederson and Salomen (1977) advocate that if accurate comparisons are to be made between brief tone physiological thresholds and conventional psychoacoustic thresholds, some allowance for the effects of temporal integration must be made. They recommend the use of a "temporal integration function" which predicts degree of temporal integration from the magnitude of the hearing loss.

Previous studies have also recognised the possible effects of decreased temporal integration and loss of frequency selectivity on ABR-behavioral comparisons.(McGee & Clemis, 1980, Kileny & Magnathan, 1987). It is clear that variables such as these whose effects are, in themselves, difficult to predict, confound further the comparison of electrophysiological and behavioural thresholds and provide another definite source of predictive error.

Two further areas of possible contention also concern phenomena associated with changes caused by cochlear damage. As mentioned in the introduction, notched noise masking has a built in error factor caused by upward spread of masking from the lower frequency edge of the notch (Stapells, 1985, Gorga, 1983). Smits and Duifhuis (1982) have shown that individuals with hearing loss can demonstrate an increased upward spread of masking. The implication is that in the present study the impaired subjects may have suffered a relatively greater effect of masking noise on the tone pips than normal subjects. Another problem relating to masking concerns the finding that hearing impaired subjects demonstrate wider critical bandwidths than normals. Critical bands are those bands of frequencies in a noise beyond which further bandwidth broadening does not increase the masking of a pure tone in the centre of the band (Bilger & Hirsch, 1956). If critical bandwidths grew to a size greater than the width of the notch in the noise used, masking would be introduced into the notch and reduce responses to the tone burst, perhaps again resulting in possible overestimation of actual hearing loss. Both upward spread of masking and critical bandwidths as they apply to tone pip ABRs in notched noise require further investigation.

Apart from standardising the response state of subjects and employing a more objective response detection technique, the inclusion of two other modifications to the procedure, might have reduced the predictive error and added to the validity of the study. The first concerns the use of non-linear stimulus gating functions to provide greater frequency specificity with rapid onset stimuli. Gorga et al (1983) note that turning tone bursts on and off with "more sophisticated gating functions" (p. 356) based on normal distribution curves or cosine squares, can significantly improve the relationship between centre and side lobes of the resultant energy spectra. Secondly, the utility of the 2-1-2 cycle stimulus at higher frequencies could have been assessed had the equipment available allowed accurate production of envelope parameters less than 1 msec in duration (Houghton, 1987). Both these two limitations apply to most clinical situations, as many commercial evoked response systems provide only limited scope for the alteration of stimulus parameters.

The applications of frequency specific ABR work are further restricted by the factor of age. Kaga and Tanaka (1980) found that neonates had the highest and adults the lowest thresholds to both standard ABR and behavioural stimuli. They reported ABR thresholds of 32 dB HL and 5 dB HL compared to behavioural thresholds of 86 dB HL and 1 dB HL for neonates and adults respectively. The important note here is that the difference between behavioural and electrophysiological measures changed significantly with age - emphasising the need for normative data in different age groups to allow accurate prediction of one measure from the other.

Lastly, all ABR work is ultimately limited by the overall validity of the response, The absence of an ABR does not mean that no hearing is present, as this can occur idiopathically or due to brainstem neurological disorders, such as demyelinating disease, and acoustic trauma (Stapells et al., 1985, Hyde et al., 1986). Conversely, the presence of an ABR does not equate with normal hearing. Jacobsen and Hyde (1985) point out that disorders of auditory pathways occurring above the site of generation of the ABR may allow a normal ABR to be obtained even when a subject has little or no functional hearing.

CONCLUSION.

In summary then, the ability of the tone pip ABR in notched and high pass noise to accurately predict behavioural hearing thresholds is variable. If a fixed difference between degree of loss and degree of ABR threshold elevation existed, the applications of such a test to both medical and educational professionals are obvious. However, our results suggest that no one such correction factor exists for subjects with cochlear hearing impairment, and that attempting to use such methods makes ones predictions prone to errors in the range of -17 to +24 dB. This range agrees closely with the findings of Stapells et al (1985) who concluded that a range of -10 to +30 dB should be allowed in the conversion from electrophysiological measures to pure tone audiogram. In many clinical situations these kinds of errors make the audiometric utility of such conversions limited.

Davis et al (1985) suggest however, that an error range of ±10 dB is adequate in the clinical setting for gross distinctions between cochlear, conductive, and central hearing losses. Looking at the overall standard deviation of 8 dB obtained in this study, the technique employed here for obtaining frequency specific estimates does appear to provide results within these limits. Hence when unable to obtain audiometric results via conventional testing, the ABR method outlined here can at least provide a reasonable approximation of both degree and configuration of any cochlear hearing loss present. As can be seen however, the accuracy of such an approximation is variable, though much of the variation evident in our results may stem from changes in the auditory system associated with cochlear pathologies. Thus while the 'accuracy' of tests such as this with normal listeners is well documented, a number of factors question the validity of generalising such normative data to hearing impaired listeners. Whether it is possible to reduce the error variation inherent in the present study's results through the modification of areas such as stimulus parameters, recording techniques, subject state standardisation or more objective judging procedures, is a question answerable only by further research.

APPENDIX 1

Contralateral Masking.

This was introduced whenever the level of the tone pips exceeded 20 dB nHL (i.e. 20 dB higher than the average behavioural threshold in normal listeners for the tone pips) in the non-test cochlea. Assuming a conservative 40 dB interaural attenuation (IA) :

500 Hz: 20 dB nHL = 20dB dial +40 dB = 60 dB dial(test ear)
1000 Hz: 20 dB nHL = 30dB dial +40 dB = 70 dB dial(test ear)
2000 Hz: 20 dB nHL = 25dB dial +40 dB = 65 dB dial(test ear)
4000 Hz: 20 dB nHL = 30dB dial +40 dB = 70 dB dial(test ear)

For a signal to noise ratio of -5 dB, (effective masking levels from Houghton 1987), and given that dial reading was 30 ± 1 dB below peSPL levels at all frequencies:

at 500 Hz: 60 dB dial = 60+30+5 = 95 dB SPL
at 1000 Hz: 70 dB dial = 70+30+5 = 105 dB SPL
at 2000 Hz: 65 dB dial = 65+30+5 = 100 dB SPL
at 4000 Hz: 70 dB dial = 70+30+5 = 105 dB SPL

From this, the following masking table was constructed:

TEST EAR (tone pip level)	NON TEST EAR (masking required)
60 dB dial=90 dB SPL	55 dB SPL = 45 dB dial
65 dB dial=95 dB SPL	60 dB SPL = 50 dB dial
.	.
.	.
90 dB dial=120 dB SPL	85 dB SPL = 75 dB dial

APPENDIX 2

Tukey Standardised Range Tests:

MS Error = 58.22

df = 81

s = 30 (subjects)

r = 4 (treatments)

Data Means:

500 Hz	1000 Hz	2000 Hz	4000 Hz
40	31	34	32

$q_t = 3.74$ $\vec{d}_t = \dfrac{3.74 \times \sqrt{58.22}}{\sqrt{30}}$

$= 5.209$

Significant results:

500 Hz - 1000 Hz = 9.0

500 Hz - 2000 Hz = 6.0

500 Hz - 4000 Hz = 8.0

Non-significant results:

1000 Hz - 2000 Hz = 3.0

1000 Hz - 4000 Hz = 1.0

2000 Hz - 4000 Hz = 2.0

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